The sensitivity of joint kinematics and kinetics to marker placement during a change of direction task

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Abstract

The conventional gait model (CGM) refers to several closely related biomechanical models used in the objective analysis of human motion. Their use has become popular in the analysis of change of direction tasks to inform best practice in the prevention and rehabilitation of anterior cruciate ligament injury. As externally-placed markers define segment axes origins and orientations, kinematic and kinetic outputs from the CGM are sensitive to marker placement. The aim of this investigation was to quantify the sensitivity of lower extremity kinematics and knee moments to systematic differences in marker placement across the stance phase of a change of direction task. Systematic anterior/posterior displacements were applied to the lateral thigh, femoral epicondyle and tibia markers in software. One-dimensional statistical parametric mapping was used to determine the effect of marker placement across the entire stance phase of a 90° change of direction task. Marker placement error within previously reported inter-tester variability ranges caused significant differences in knee abduction moment, hip rotation angle, knee rotation angle, ankle rotation angle and ankle abduction angle across various periods of stance. Discrete measures of these variables have been associated with increased frontal plane knee loading during change of direction, considered a key mechanism of anterior cruciate ligament injury. Systematic differences in marker placement may lead to incorrect group statistical inferences in such discrete measures.

Introduction

The conventional gait model (CGM) refers to several closely related biomechanical models, the data from which are used to analyse human motion, inform clinical decision making and evaluate rehabilitation interventions (Baker et al. 2017). Such models provide an objective record of kinematic and kinetic metrics during movement. Originally developed for and implemented in clinical gait analyses, the CGM's application has been extended to a variety of movements, including a range of change of direction (CoD) tasks (Franklyn-Miller et al. 2017; King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, et al. 2018; B. M. Marshall et al. 2014; McLean, Huang, and Van Den Bogert 2005; O'Malley et al. 2018; Sigward and Powers 2007).

CoD is the most common mechanism of non-contact anterior cruciate ligament (ACL) rupture, a serious musculoskeletal injury normally requiring surgical intervention (Kvist 2004). The CGM has been utilised in the analysis of CoD to inform best practice in the prevention and rehabilitation of ACL injury (King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, et al. 2018; McLean, Huang, and Van Den Bogert 2005; Sigward and Powers 2007). Kinematic variables at the hip, knee and ankle have been associated with increased frontal plane knee loading during CoD, considered a key risk factor for injury (Hewett et al. 2005; McLean, Huang, and Van Den Bogert 2005; Sigward and Powers 2007).

Accurate measures of these variables rely on the correct definition of body segment axes origins and orientations (Kadaba et al. 1989). In the Plug-in-Gait (PiG) model (Vicon, Oxford Metrics, London, UK), a widely used implementation of the CGM, retroreflective markers placed externally on a series of anatomical landmarks define

segment origins and orientations. Variation in marker placement is cited as the primary factor in the low reliability indices reported for many kinematic and kinetic variables (Alenezi et al. 2016; Gorton, Hebert, and Gannotti 2009; McGinley et al. 2009).

Inter-tester variability in anatomical landmark location, and subsequently marker placement, makes inferring ACL injury mechanisms based on data collected in different laboratories and by different practitioners challenging. The range of inter-tester variability in anatomical landmark location for marker positions has been reported as 12 – 25 mm (Della Croce, Cappozzo, and Kerrigan 1999). Given their roles in defining the origins and orientations of the femur and shank segments, the lateral thigh (THI), lateral femoral epicondyle (KNEE) and lateral tibia (TIB) markers have the largest effect on model outputs (Kadaba, Ramakrishnan, and Wooten 1989). The deterministic nature of the model indicates that variation in the anterior/posterior positions of these markers will alter joint kinematics and kinetics at the hip, knee and ankle (Kadaba, Ramakrishnan, and Wooten 1989).

Experimental studies confirm the sensitivity of joint kinematics, particularly frontal and transverse plane kinematics, to marker placement error during walking (Baker, Finney, and Orr 1999b; Ferrari et al. 2008; Groen et al. 2012; Kadaba et al. 1989; Szcserbik and Kalinowska 2014). Simulated displacements in THI marker position cause large errors in transverse plane hip and frontal plane knee kinematics, both of which have been associated with increased frontal plane knee loading during CoD (Baker, Finney, and Orr 1999b; McLean, Huang, and Van Den Bogert 2005; Sigward and Powers 2007). Errors in frontal plane knee kinematics vary non-uniformly throughout the gait cycle, demonstrating analysis of the entire gait cycle may be required to fully understand the effect of marker placement on joint kinematics.

Calculated joint moments of force are also affected by marker placement. Changing the positions of the THI, KNEE and TIB markers alters the locations of the calculated knee (KJC) and ankle joint centres (AJC), affecting the length of the moment arm used to calculate the joint moment. Simulated displacements in joint centre positions demonstrate this, with 10 mm anterior displacements causing significant differences in net knee moments during walking (Holden and Stanhope 1998; Stagni et al. 2000).

The specific sensitivity of kinematic and kinetic variables to systematic differences in marker placement remains unclear. The effect of marker placement will vary depending on the variable being reported, the marker in question, the magnitude of displacement and the phase of the movement being analysed. To reliably make inferences related to ACL injury from data collected in different laboratories and by different practitioners, we must establish the sensitivity of lower extremity kinematics and knee moments to systematic differences in marker placement. The aim of this investigation was to determine the sensitivity of joint kinematics at the hip, knee and ankle, as well as knee moments, to systematic displacements in the positions of the THI, KNEE and TIB markers across the stance phase of a CoD task.

Methods

Participants

An *a priori* power analysis (G*Power, version 3.1.9.2, Universität Düsseldorf, Germany), based on previously published data (Alenezi et al. 2016), indicated that a sample size of 42 participants was required to achieve 80% statistical power with an alpha level of 0.05. Fifty eligible participants (mean ± SD: 24.8 ± 4.8 years, 180 ± 6

cm and 84 ± 15.3 kg) were consecutively recruited from the caseload of two orthopaedic surgeons based in the Sports Surgery Clinic, Dublin, Ireland.

Inclusion criteria for participation were: male, aged 18 – 35, undergone primary ACLR 34 – 43 weeks (mean ± SD: 35.7 ± 1.2 weeks) prior to testing, participation in multi-directional field-based sport prior to ACL injury and intention to return to the same level of participation following rehabilitation. The study received ethical approval from the University of Roehampton, London (LSC 15/122) and the Sports Surgery Clinical Hospital Ethics committee (25AFM010). Participants gave informed, written consent prior to participation in the study.

Data Collection

Testing took place in a biomechanics laboratory, using a ten-camera motion analysis system (200 Hz; Bonita-B10, Vicon, UK), synchronized (Vicon Nexus 2.7) with two force platforms (1000 Hz BP400600, AMTI, USA) recording the positions of 28 reflective markers (14 mm diameter). Markers were secured to the participant's shoe or skin using tape at bony landmarks on the lower limbs, pelvis and trunk according to the PiG marker set (B. M. Marshall et al. 2014).

Prior to data collection, participants undertook a standardised warm-up comprising of a 2-minute jog, 5 bodyweight squats, 2 submaximal and 3 maximal countermovement jumps. A static trial was captured as a reference for the dynamic trials. Each participant completed a pre-planned 90° CoD task. The CoD task followed a wider testing battery that formed part of a larger, ongoing study, in which participants also completed a range of double and single leg jump exercises. The CoD task involved the participants running maximally towards the force platforms then planting their outside foot on the force platform to cut left or right, i.e. planting

their left foot to cut to the right. Three valid, maximal effort trials were collected on both the non-operated and operated limb. A full description of the testing protocol is given in King et al. (2018).

Data Processing

Trials in which the participant planted their operated limb on the force platform to complete the CoD task were used for further analysis. Marker trajectory and force data were low-pass filtered using a fourth-order Butterworth filter (cut-off frequency 15 Hz) (Kristianslund, Krosshaug, and Bogert 2012). Systematic displacements were then applied in software to the positions of the THI, KNEE and TIB markers. One marker position displacement was applied at a time along the corresponding segment *x*-axis using

$$X_k' = T.X_k$$

where X_k' are the new, displaced marker coordinates within the segment coordinate system, T is the translational matrix and X_k are the original marker coordinates within the segment coordinate system (Fig 1). Displacements were applied to marker positions in 5 mm increments, to 20 mm anterior and 20 mm posterior from their original positions, resulting in 8 displacement conditions for each marker. Data processing created three separate data sets: A, B and C. Each data set contained displacements of a single marker and were identical except for the position of the corresponding marker.

Stance phase was identified for each trial from when vertical ground reaction force passed above and below 20 N. Tri-planar joint angles at the hip, knee and ankle, as well as tri-planar knee moments were extracted during stance phase for each trial.

Kinematic and kinetic signals were time normalised to 101 data points and the mean of each participant's three trials was used for further analysis.

Sensitivity Analysis

One-dimensional statistical parametric mapping (SPM) was used to analyse the effect of marker placement across the entire stance phase of the CoD task (Pataky 2010, 2014; Pataky, Robinson, and Vanrenterghem 2013). Our analysis aimed to simulate a scenario in which we were testing for between group differences in groups which were identical except for the position of the corresponding marker. This would allow us to identify the minimum systematic differences in marker placement required to result in incorrect statistical inferences when making between group comparisons in each variable. For clarity, we will use the example of one data set, data set A, as the process was repeated identically for data sets B and C. Following data processing, nine signals for each variable for each participant were contained in data set A. These corresponded to the original unaltered trial, as well as each of the THI marker displacement conditions (Fig 3).

Each variable in data set A was submitted to a 1D independent samples SPM t-test between the unaltered condition and each of the displacement conditions. This process produced 8 SPM{t} curves for each variable, one for each THI marker displacement condition (Fig 4). The significance of each SPM{t} curve was determined topologically using random field theory ($a < 0.05$) (Pataky, Vanrenterghem, and Robinson 2015). Phases of the SPM{t} curve above the critical-t threshold were identified as significantly affected by the corresponding marker displacement. To aid in interpretation of results, SPM{t} curves were plotted using

188 image inference surface plots (Fig. 5). A variable's "sensitivity" to marker placement
189 was determined by the minimum marker displacement required to cause significant
190 differences, with more sensitive variables significantly affected by smaller marker
191 displacements across larger periods of stance phase.

192 As we experimentally created the difference between conditions by displacing each
193 marker in a fixed direction from its original position, the changes to outcome
194 variables will be unidirectional and predictable in nature. For example, an anterior
195 displacement of the THI marker will always result in a more internally-rotated
196 calculated position of the thigh segment. The test statistic produced following
197 comparisons between the unaltered condition and each displacement condition is
198 therefore a function of sample size and effect size, meaning that the likelihood of
199 finding a statistically significant differences between conditions is increased at larger
200 sample sizes. In acknowledgment of this, we included sample size as an extra
201 degree of freedom in our analysis. We chose sample sizes of n = 10, n = 25 and n =
202 50, as these represent the low, mid and upper ranges of sample sizes typically used
203 in biomechanical studies (Besier, Lloyd, and Ackland 2003; Ithurburn et al. 2017;
204 Sankey et al. 2015; Wen et al. 2018). The sensitivity analysis procedure outlined
205 above was repeated for each variable in data sets A, B and C, at each sample size,
206 resulting in a total of nine sensitivity analyses.

207 **Results**

208 The results of the sensitivity analyses for the THI, KNEE and TIB markers are
209 presented in Figures 6, 7 and 8 respectively. See supplementary material –
210 Appendix A, for individual sensitivity analyses for each variable. As sample size
211 increased, the magnitude of the marker displacement required to cause significant

differences in each variable decreased, and/or the cumulative percentage of stance phase significantly affected by marker displacements increased.

Thigh Marker

No variables were significantly affected by 5mm THI marker displacements. Four variables were significantly affected by displacements of 10 mm and greater across periods of early, mid and late stance (Fig 5B, 6C). These variables were hip rotation angle, knee abduction angle, ankle abduction angle and ankle rotation angle. Of these, hip rotation and knee abduction angles were most sensitive to THI marker placement, with 10 mm displacements causing significant differences across the entire stance phase at n = 50 (Fig 5C). At n = 10, only hip rotation and knee abduction angles were significantly affected by THI marker displacements of any magnitude. The sensitivity of these variables increased as sample size increased, while at n = 25 and n = 50, ankle abduction and rotation angles were also significantly affected (Fig 5B, 5C).

Knee Marker

No variables were significantly affected by 5 mm KNEE marker displacements (Fig 6). Eight variables were significantly affected by KNEE marker displacements of 10 mm and above (Fig 6C). These were hip rotation angle, knee flexion angle, knee rotation angle, ankle plantar-flexion angle, ankle abduction angle, knee flexor moment and knee abduction moment (Fig 6B, 6C). Of these, ankle abduction and rotation angles were most sensitive to KNEE marker displacements, with 10 mm displacements causing significant differences across the first and last 20% of stance (Fig 6C). At n = 10, no variables were significantly affected by KNEE marker displacements of any magnitudes. At n = 25, ankle plantar-flexion, ankle abduction,

ankle rotation, knee flexor moment and knee abduction moment were significantly affected (Fig 6B), while at n = 50, hip rotation, knee flexion, knee abduction and knee rotation angles were also significantly affected (Fig 6C).

Tibia Marker

5 mm TIB marker displacements significantly affected three kinematic variables (Fig 7C). These were, knee rotation angle, ankle abduction angle and ankle rotation angle. Displacements of 10 mm and above also significantly affected ankle plantar-flexion angle, knee flexor moment and knee abduction moment (Fig 7B, 7C). Knee rotation angle was the most sensitive variable to TIB marker displacements, and the only variable to be significantly affected across the entire stance phase by any 5 mm marker displacements (Fig 7C). At n = 10, knee rotation angle, ankle abduction angle, ankle rotation angle and knee abduction moment were significantly affected by TIB marker displacements (Fig 7C). The sensitivity of these variables increased as sample size increased, while ankle plantar-flexion angle and knee abduction moment were also significantly affected at n = 25 and n = 50 (Fig 7B, 7C).

Discussion

Inter-tester variability in the anterior/posterior positions of the anatomical landmarks used to define the positions of the THI, KNEE and TIB markers is reported as ranging between 9.3 – 12.5 mm (Della Croce, Cappozzo, and Kerrigan 1999). Several variables previously associated with ACL injury risk and rehabilitation status were significantly affected by marker displacements within, or bordering on, reported inter-tester variability ranges. These were hip rotation angle, knee abduction angle,

ankle rotation angle and knee abduction moment (Dempsey et al. 2007; McLean, Huang, and Van Den Bogert 2005; Sigward and Powers 2007).

Frontal and transverse plane kinematics were most sensitive to marker placement in each marker condition and at every sample size. This is unsurprising given the known limitations of the CGM in assessing frontal and transverse plane kinematics (Baker, Finney, and Orr 1999a; Kadaba, Ramakrishnan, and Wooten 1989). Changes in the anterior/posterior positions of the THI, KNEE and TIB markers causes misalignment of the primary and secondary axis of the femur and shank segments. These alterations create a rotational offset, while also resulting in cross-talk between segment axes. This manifests as error in angles calculated in all three planes, and is most pronounced in the frontal and transverse plane kinematics (Baker, Finney, and Orr 1999b). Previous studies using descriptive statistics (Szczerbik and Kalinowska 2011), root mean square differences (Groen et al. 2012) and qualitative assessments (Kadaba et al. 1989) to examine the effect of marker placement on joint kinematics during walking report similar findings.

Our findings build on those from previous work and demonstrate the minimum systematic differences in marker placement required to cause statistically significant differences in each variable at three different sample sizes. Utilising a continuous statistical analysis method (SPM) allowed us to identify the specific phases of each kinematic and kinetic signal significantly affected by marker displacements. Statistically significant differences first appeared in many outcome variables across the first and last 20% of stance, indicating these phases are most sensitive to marker placement (Fig 5A, 6B, 7A). As non-contact ACL injuries are believed to occur within the first 20% of stance, discrete kinematic and kinetic measures from this period are regularly reported (Pollard, Sigward, and Powers 2007a; Sigward and Powers 2007;

Stearns and Pollard 2013). Increased hip internal rotation, knee abduction and ankle external rotation at initial contact of CoD have been associated with higher peak knee abduction moments (Dempsey et al. 2007; McLean, Huang, and Van Den Bogert 2005; Sigward and Powers 2007). Frontal plane knee loading is considered a key risk factor for ACL injury (Hewett et al. 2005). These findings have thus led to the clinical development of ACL prevention and rehabilitation programs aiming to minimise frontal plane knee loading (Distefano et al. 2011).

Statistical significance is often used to draw clinical inferences in ACL research (Dempsey et al. 2007; Ford et al. 2005; King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, et al. 2018; Sigward and Powers 2007; Stearns and Pollard 2013). Previous work has reported statistically significant differences in kinematics and kinetics with respect to gender (Ford et al. 2005), limbs (King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, et al. 2018) and injured/uninjured groups (Stearns and Pollard 2013) and postulated that these differences may highlight variables of interest in rehabilitation and injury prevention. It should be noted that statistical significance is less relevant than the actual magnitude of differences between groups and how such differences would affect clinical inferences/recommendations. Relative to previously published differences, our findings demonstrate magnitudes approximating or exceeding those reported between groups/conditions (Ford et al. 2005; King, Richter, Franklyn-Miller, Daniels, Wadey, Jackson, et al. 2018; Pollard, Sigward, and Powers 2007b; Stearns and Pollard 2013). For example, statistically significant differences in hip rotation angle (5.1°), knee abduction angle (2°) and knee abduction moment (0.21, 0.53 and 1 Nm/kg) during CoD tasks have been reported previously and hypothesised to present clinically relevant differences related to ACL injury (McLean, Huang, and Van Den Bogert 2005; Sigward and

310 Powers 2007; Stearns and Pollard 2013). Within our data, at n = 50 10 mm THI
311 marker displacements caused significant differences in hip rotation and knee
312 abduction angle with a mean difference of 3.62° and 2.77° respectively, while 10 mm
313 TIB marker displacements caused significant differences in knee abduction moment
314 with a mean difference of 3.22 Nm/kg (see supplementary material – Appendix A).

315

316 Several limitations can be ascribed to the current study. Firstly, we do not know if the
317 original physical marker positions were optimal. Moving the markers
318 anteriorly/posteriorly may have in fact been moving them closer to the original target
319 positions. However, as the effect of systematic marker displacements on outcome
320 variables is unidirectional, the original marker locations will not affect our general
321 conclusions. Secondly, there is there is likely to be an element of random variation in
322 real-world marker placement, alongside the systematic element investigated here
323 (Osis et al. 2016). Random marker placement error and its effect on kinematics and
324 kinetics requires further research**.** Also, it is important to note that the specific errors
325 reported in this study are limited to the CoD task analysed, with marker placement
326 likely having a different effect in different tasks (Baker, Finney, and Orr 1999a).
327 Lastly, our marker displacements were simplistic in nature and do not directly mimic
328 real world marker placement error. We implemented fixed displacements, meaning
329 markers were moved the same distance relative to the original marker position
330 across all time points of the task. Physically moving markers across a range of ± 20
331 mm on the skin would involve a certain amount of medio-lateral in addition to
332 anterior/posterior displacement, as well as different soft tissue artefacts (STA).
333 Different STA's would alter the observed errors in this study, meaning translating our
334 findings directly to real world scenarios is challenging. Separating the effect of

marker placement error from that of STA is difficult and the relationship between these two major sources of error is an area that warrants further research. For this study, we chose to focus on simple anterior/posterior displacements, as the model definitions indicate that these are the marker displacements that most substantially effect model outputs (Kadaba, Ramakrishnan, and Wooten 1989). Accounting for the additional effects of medio-lateral displacements and STA went beyond the scope of the current investigation.

Alternative methods for modelling the human body have been developed to mitigate the effect of STA and provide improved anatomical relevance compared to the CGM. These include models that implement the calibration anatomical systems technique (CAST), or models that allow for six degrees of freedom (6DOF) at each joint. Models implementing CAST or 6DOF continue to work on the assumption that marker placement is consistent and repeatable between practitioners (Charlton et al. 2004). Indeed, any model utilising anatomical markers to define joint centres and segment orientations makes this assumption. At present no alternative model or technique has been as widely implemented and validated as the CGM (Baker et al. 2017; Charlton et al. 2004). Research into the sensitivity of alternative modelling techniques to marker placement, and how this compares to the CGM is required prior to any widespread clinical application. While limited in certain aspects, the CGM currently presents a practical, deterministic, extensively validated model that can be easily implemented in routine clinical practice. These factors may explain the continued widespread use of the CGM in contemporary biomechanical research (Cortes, Onate, and van Lunen 2011; Gore et al. 2018; Lee, Chow, and Tillman 2014; B. Marshall et al. 2015; McLean, Huang, and Van Den Bogert 2005; Pollard, Sigward, and Powers 2007a; Sigward and Powers 2007). When utilising the CGM

however, it should be done in a manner that openly acknowledges its limitations within the context of the study aims and reported results. If attempting to identify relatively small differences in frontal and transverse plane kinematics for example, it should be made explicitly clear that any identified differences may be attributable to instrumental error such as marker placement.

In conclusion, we have shown that systematic differences in the placement of the THI, KNEE and TIB markers, within or bordering on reported inter-tester variability ranges, can cause statistically significant differences in multiple kinematic and kinetic variables across various periods of CoD stance. Many variables affected have previously been associated with increased frontal plane knee loading during CoD, which is considered a key risk factor for ACL injury. Errors were particularly pronounced across the first 20% of stance, a period from which discrete kinematic and kinetic variables are regularly reported. Our findings demonstrate the minimum systematic differences in marker positions required to cause significant differences in lower extremity kinematics and kinetics. These thresholds can be used by laboratories to establish acceptable levels of inter-tester variability in marker placement. If inter-tester variability is above these thresholds, statistical inferences and corresponding clinical recommendations related to group differences should be made with caution, as marker placement differences may result in invalid conclusions.

Conflict of interest statement

The authors confirm that there is no financial or personal relationship with other individuals or organisations that could inappropriately influence this work.

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